

"Method and apparatus for fibred high-resolution fluorescence imaging, in particular confocal imaging"

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The present invention relates to a method and apparatus for fibred
5 high-resolution fluorescence imaging, in particular confocal imaging. The intended field of application is more particularly that of *in vivo* and *in situ* imaging.

The fluorescence observed may come from an exogenous compound (typically an injected marker) or an endogenous compound (present in the
10 cell) of a biological tissue.

More particularly, the method of confocal imaging according to the invention is of the type which consists of scanning a tissue point by point in a subsurface plane, each point corresponding to an excitation signal emitted by a continuous source, deflected and injected by turns into an optical fibre of a
15 bundle of optical fibres, then focussed at the exit of said fibre in said plane, each point emitting in return a fluorescence signal collected by said optical fibre, then detected and digitized to form an image element.

The confocal character is obtained using the same optical path, in particular the same optical fibre which is used for spatial filtering, to transport
20 the excitation signal and the fluorescence signal emitted in response, and using an optical system adapted to conjugate the focussing point in the tissue with said optical fibre.

The corresponding confocal imaging apparatus comprises the bundle of flexible optical fibres (usually called image guide) with, at its proximal end:
25 - the source emitting continuously or in pulsed mode at the excitation wavelength of one or more targeted fluorophores, typically a laser source;
 - means for rapid scanning over time of the excitation beam produced by the source in lines and columns in a XY plane corresponding to the entry section of the image guide;

- means for injecting the excitation beam into one of the optical fibres;
- means for detecting the fluorescence signal; and
- control means, in particular for the scanning means.

5 Means are also provided adapted for the realization and the display of an image from the signals successively detected on each fibre.

At the distal end of the image guide is an optical head, intended to be brought into contact with the observed tissue, allowing the excitation beam to be focussed at a given depth (tens of μm) and thus the realization of a
10 subsurface image.

In practice, this type of apparatus has the following advantages in particular:

- on the distal side of the image guide there is only one focussing optics, which is less fragile and less costly than an optical head with incorporated
15 scanning means, the replacement of the optical head being able to be envisaged independently of the scanning means; moreover the head can be miniaturized, which is advantageous in endoscopy and also, in a general way, for increasing the precision of positioning;

- an image guide made of flexible optical fibres, serves as an access arm to
20 the site to be observed, which is important for an *in situ* application.

In the state of the art, the following fluorescence methods and apparatus have been described.

In an article in "Applied Optics", Vol. 38, No. 34, December 1999, pages 7133-7144, is presented a confocal microendoscope using a bundle of
25 flexible optical fibres and equipped at its distal end with a miniature optical focussing head; it is provided to scan the bundle of optical fibres in lines, the apparatus comprising a detection slit as well as a CCD linear detector of the charge-transfer type. With an apparatus of this type 4 images/s can be obtained, which is too slow for *in situ* imaging dependent on the movement

of the subject and of the operator. Moreover, the fact of scanning in lines and not point by point degrades the confocal character and leads to a less well resolved image.

In an article in "Optics Communication", 188 (2001), pages 267-273, is
5 presented the coupling of a confocal laser scanning desktop microscope with
a bundle of flexible optical fibres which is equipped at its distal end with a
miniature optical head. The desired aim is to make the microscope compatible
with an endoscope. The scanning is carried out fibre by fibre but the desktop
microscope used is of standard design, designed to display a fixed specimen
10 without concern for the time taken to obtain an image. An exposure time of 2
seconds is proposed in this article, much too long for a real-time *in situ*
imaging.

The present invention aims to propose a method and apparatus for
fibred high-resolution fluorescence imaging, in particular confocal imaging, of
15 the type using an image guide made of optical fibres scanned one by one by
an excitation beam emitted continuously, said method and apparatus making
it possible to obtain the display in real time of a site *in vivo in situ*, i.e. capable
of providing a sufficient number of point by point images per second without
being dependent on the movements of the subject and of the operator in order
20 to obtain in particular a fairly rapid examination.

An aim of the present invention is also to propose a method and
apparatus which, in a general way, optimize the quality of each image in order
to obtain in particular an excellent lateral and axial resolution.

According to a first feature of the invention, a method is proposed for
25 the realization of a confocal fluorescence *in vivo in situ* image, the method
using an image guide made of several thousands of optical fibres and
consisting of the point by point scanning of a tissue in a subsurface plane,
each point corresponding to an excitation signal emitted by a continuous
source, deflected and injected into one of the optical fibres of said bundle then
30 focussed on the exit of said fibre in said plane, each point emitting in return a
fluorescence signal collected by said optical fibre, then

detected and digitized to form an image element, characterized in that the excitation signal is deflected at a speed corresponding to acquisition of a number of images per second sufficient for a real-time use and in that the fluorescence signal is detected at a detection frequency corresponding to a
5 minimum sampling frequency of the fibres one-by-one.

According to a second feature of the invention, a method is proposed for the realization of a high-resolution fluorescence image using an image guide made of several thousands of optical fibres, an excitation signal being emitted by a continuous source, deflected and injected by turns into one of the
10 optical fibres of said image guide and a fluorescence signal emitted in response being collected by the same optical fibre as the one used for the excitation, then detected and digitized to form an image element, characterized in that the end of the fibres is intended to be placed bare directly in contact with the surface of the tissue to be imaged, each fibre being
15 adapted to produce a divergent beam which is adapted to excite a microvolume of the tissue situated at the surface to a maximum depth depending in particular on the core diameter of the optical fibres and in that the excitation signal is deflected at a speed corresponding to acquisition of a number of images per second sufficient for a real-time use and in that the
20 fluorescence signal is detected at a detection frequency corresponding to a minimum sampling frequency of the fibres one-by-one.

This method thus differs from the method according to the first feature in that it does not provide the scanning of a signal which is focussed on the exit of each fibre but the scanning of a divergent signal on the exit of each
25 fibre. The non-focussing of the signal on the exit of the fibre makes it possible to obtain images of a volume situated just beneath the surface of the tissue, images which can be processed and which are advantageous in particular from a medical point of view. These images are not "confocal" because they do not originate from a subsurface section plane that has been scanned point
30 by point, but images which can nevertheless be called "high-resolution" because they come from the scanning by turns of

microvolumes situated directly beneath the surface and from a spatial filtering of the fluorescence signal emitted by each microvolume by the same fibre as the one which was used for the excitation. The main advantage of such an apparatus resides in the fact, that for an endoscopic application, the diameter
5 of the endoscopic probe can be very small depending solely on the diameter of the image guide and therefore on its number of optical fibres. This allows fields of application to be envisaged, such as for example the field of neurology, where the size of the endoscopic probe is a critical factor in overcoming the problems inherent in the miniaturization of the optical
10 focussing head.

The present invention proposes a solution for realizing an in vivo in situ image in real time, whether for a method with or without focussing on the exit of the fibre. The invention is based on respecting the sampling of the fibres (according to Shannon's criterion) which makes it possible to obtain an image
15 reconstructed point by point which effectively corresponds to each fibre. This avoids loss of information when sampling all of the fibres one by one while still maintaining a minimum average number of images per second, i.e. in practice at least 12 images per second for a maximum mode of 640 x 640 pixels. The choice of detection frequency (pass-band of the detector) as a function of this
20 minimum sampling then allows the detection, for each fibre, of the greatest possible number of fluorescence photons.

Thus, according to a possible embodiment, using an image guide with approximately 30 000 flexible optical fibres, the sampling frequency and the pass-band of the detection system (an avalanche photodiode or equivalent)
25 are fixed at approximately 1.5 MHz, corresponding approximately to 12 pixels per fibre, thus allowing as a minimum the 12 images/s in maximum mode 640 x 640 pixels to be obtained.

In practice, according to an advantageous embodiment, which makes it possible an appropriate rapid scanning of the fibres in order to obtain a real-time image, the deflection of the beam is adjusted by determining a rapid-resonance frequency
30

of a resonating "line" mirror and a slow-resonance frequency of a galvanometric "frame" mirror.

According to the invention, the detection period for a flux coming out from a fibre is time-limited and is very short to respect the acquisition of a real-time image. According to the invention, optimization means have thus also been provided to collect and detect the maximum flux coming from the specimen during this limited detection period, in particular:

- optical deflection, injection, if appropriate focussing and detection means are used presenting a degree of achromaticity which depends on the wavelength difference between the excitation wavelength and the maximum fluorescence wavelength, as well as on the spectral width of the emitted fluorescence signal; this advantageously allows optimization of the fluorescence flux collected over the whole of the spectral band of emitted fluorescence;

- if appropriate, a numerical aperture of the focussing optics on the tissue comprised between approximately 0.5 and 1 is chosen, which makes it possible to achieve a good compromise between the field size and the quantity of photons collected; and

- a detector is chosen having a quantum efficiency at the fluorescence wavelengths to be detected of at least 50 %.

According to the invention, the image processing then carried out on the detected flux is also optimized, in order to obtain an image of very good quality from this limited flux of detected photons. This optimization is carried out in the following way.

A series of steps is carried out prior to the acquisition of images in real time:

- a step of detection of the location of each fibre of a chosen set of fibres intended to be used (either all of the image guide or a chosen sub-set); this step is to be carried out at least at each change of image guide;

- a step of calibrating the rate of injection into each fibre, i.e. of defining a rate of injection particular to each fibre; and
- a step of detection of the background image (without specimen).

5 During operation, the optimization of the image processing comprises in particular the steps consisting, after digitization of the detected signal, of:

10 - defining the real flux collected by each fibre, i.e. only coming from the specimen, after correction as a function of the rate of injection particular to the fibre and subtracting the background image, so as to obtain a corrected signal;

- then undertaking a reconstruction of the image from this corrected signal, in particular with the aim of transforming an image showing a mosaic of fibres in an image without apparent fibres.

According to the invention, these last two steps must be able to be carried out in real time. As regards the correction of the signal, this can take place in real time thanks to a processing which is suited to the structure of the observed signal and an optimized algorithm. As regards the reconstruction of the image, this can take place by choosing a number of operations per pixel which can be carried out in real time allowing the desired result to be achieved 20 in terms of image quality. A Gaussian low-pass filtering represents a good compromise between the complexity of the processing, the quality of the result and the calculation time.

In order to save time and to increase the complexity of the processing corresponding to the reconstruction of the image, the processing capacity of 25 the equipment can also be increased, for example by using a specific processing card and/or a parallel architecture such as a multiprocessor.

The present invention also proposes an apparatus adapted for the implementation of the method according to the first feature recited above, comprising:

30 - the image guide;

- the source emitting continuously at the excitation wavelength of at least one targeted fluorophore,
 - means for rapid scanning and for injection fibre by fibre over time of the
- 5 excitation beam produced by the source by lines and by columns in a XY plane corresponding to the entry section of the image guide;
- means for separating the excitation wavelength and the fluorescence wavelengths;
 - means for detection of the fluorescence signal; and
- 10 - means for processing the detected signal allowing the realization of an image;
- an optical head being arranged at the distal end, intended to be brought into contact with the observed tissue, allowing the excitation beam to be focussed.

The apparatus is characterized in that:

- 15 - the scanning means are suitable for moving the excitation beam at a speed corresponding to the obtaining of an image in real time; and
- the detection means have a pass-band whose frequency is fixed as a function of the minimum sampling frequency of the fibres one-by-one.

The present invention also proposes an apparatus for the

20 implementation of the method according to the second feature, comprising:

- the image guide;
 - the source emitting continuously at the excitation wavelength of at least one targeted fluorophore,
 - means for scanning and fibre-by-fibre injection of the excitation beam
- 25 produced by the source in a XY plane corresponding to the entry section of the image guide;
- means for separating the excitation wavelength and the fluorescence wavelengths;
 - means for detecting the fluorescence signal; and

- means for processing the detected signal allowing the realization of an image;

characterized in that the end of each fibre is suitable for producing a beam

5 which is divergent and is intended to be placed bare directly in contact with the surface of the tissue to be observed;

and in that the scanning means are suitable for moving the excitation beam at a speed corresponding to the obtaining of an image in real time and the detection means have a pass-band whose frequency is fixed as a function of

10 the minimum sampling frequency of the fibres one-by-one.

The present invention will be better understood and other advantages will become apparent in the light of the following description of an embodiment, with reference to the attached drawings in which:

- Figure 1 schematically represents an embodiment of apparatus for confocal fluorescence imaging including a focussing head;

15 - Figure 2 is a block diagram of the apparatus of Figure 1;

- Figure 3 schematically represents an embodiment of apparatus for high-resolution fluorescence imaging; and

- Figure 4 is a block diagram of the apparatus of Figure 3.

20 According to the example chosen and represented in Figures 1 and 2, the apparatus comprises:

- a light source 1;

- means 2 for shaping the excitation beam;

- means 3 for separating wavelengths;

25 - scanning means 4;

- beam injection means 5;

- an image guide 6 made up of flexible optical fibres;

- an optical focussing head 7;

- means 8 for rejecting the excitation beam;

30 - means 9 for focussing the fluorescence signal;

- means 10 for spatial filtering of the fluorescence signal;

- means 11 for detecting the fluorescence signal; and
- means 12 for electronic and computer processing of the detected fluorescence signal and for display.

5 These different elements are detailed below.

The light source 1 is a laser emitting at an excitation wavelength allowing a wide range of fluorophores to be excited, for example 488 nm. In order to optimize the injection into one of the fibres of the image guide 6, the excitation beam is circular in order to be able to inject a fibre with a section
10 which is also circular and, in order to optimize the rate of injection, the laser is preferably of the longitudinal monomode type in order to present the best possible wave front for the injection into a slightly multimode optical fibre. The laser emits continuously or in pulsed mode, and in a stable way (the smallest possible noise, <1 %). The available output power is of the order of 20 mW.
15 By way of example, a quantum well laser (VCSEL), a diode-pumped solid-state laser, a laser diode or a gas, such as argon, laser can be used.

The means 2 for shaping the excitation laser beam are placed on the exit of the source 1. They are made up of an afocal optical system of a magnification other than 1, comprising two lenses L1 and L2 which are
20 adapted to modify the diameter of the laser beam. The magnification is calculated so that the diameter of the beam is suitable for the means 5 of injection into a fibre.

The shaped excitation laser beam is then directed towards the means 3 provided for separating the excitation and fluorescence wavelengths. It is for
25 example a dichroic filter with a transmission efficiency of 98 to 99 % of the excitation wavelength and which thus reflects substantially the other wavelengths. The fluorescence signal, which on the return follows the same optical path as the excitation signal (confocal character), will thus be almost totally sent towards the detection path (8-11). The rejection means 8 placed
30 on the detection path serve to totally eliminate the 1 to 2 % of interfering reflections at the excitation wavelength 488 nm which pass towards the detection path

(for example a rejection filter at 488 nm, a band-pass filter allowing a transmission for example only between 500 and 600 nm, or a high-pass filter which allows a transmission above 500 nm).

The scanning means 4 then pick up the excitation beam. According to
5 the example chosen and represented in Figure 1, these means comprise a mirror M1 resonating at 4 KHz which serves to deflect the beam horizontally and thus to realize the lines of the image, a mirror M2 which is galvanometric at 15 Hz which serves to deflect the beam vertically and thus to realize the frame of the image; and two afocal systems with unitary magnification, AF1
10 situated between the two mirrors and AF2 situated after the mirror M2, these afocal systems being used to conjugate the rotation planes of the two mirrors M1 and M2 with the injection plane in one of the fibres. According to the invention, the scanning speed is determined in order to allow a real-time in vivo in situ observation of the tissues. For this, the scanning must be fast
15 enough for there to be at least 12 images displayed on the screen for a display mode of 640 x 640 pixels corresponding to the slowest mode. For display modes with fewer pixels, the number of images acquired per second will thus always be greater than 12 images/s. By way of a variant, the scanning means can in particular comprise a rotating mirror, integrated
20 components of the MEMs type (X and Y scanning mirrors), or an acousto-optical system.

The excitation beam deflected on the exit of the scanning means is directed towards the optical means 5 in order to be injected into one of the fibres of the image guide 6. These means 5 are in this case made up of two
25 optical units E1 and E2. The first optical unit E1 is adapted for correction of part of the optical aberrations at the edge of the field of the scanning means 4, the injection thus being optimized over the whole optical field (at the centre as well as at the edge). The second optical unit E2 is intended to carry out the actual injection. Its focal length and its numerical aperture were chosen to
30 optimize the rate of injection into the optical fibres of the guide 6. According to one embodiment which makes it possible to obtain the achromaticity criterion, the first unit E1 is made up of a doublet of

achromatic lenses, and the second unit E2 of two doublets of achromatic lenses followed by a lens situated close to the image guide. By way of a variant, this injection optic could be made up of any other type of standard optics, such as for example two triplets, or index gradient lenses or a 5 microscope lens (which is more costly however).

The image guide 6 is made up of a very large number of flexible optical fibres (several tens of thousands), for example 30 000 fibres of 2 μm in diameter and spaced 3.3 μm apart. In practice, the whole set of fibres of the image guide, or a chosen sub-set of these fibres, for example centred, can be 10 used.

On the exit of the optical fibre, the excitation laser beam is focussed by the optical head 7 in the specimen 13 to a point 14 situated at a given depth situated between some tens of μm and about one hundred μm , relative to the surface 15 of the specimen with which the optical head is intended to be 15 placed in contact. This depth can be for example 40 μm . The optical head thus allows focussing of the flux exiting the image guide in the specimen, but also collection of the fluorescence flux returning from the specimen. The optical head has a magnification of 2.4 and a numerical aperture on the specimen of 0.5. These two parameters are chosen so that the return signal 20 occurs only in the optical fibre which has transmitted the excitation signal and not in adjacent fibres and to thus preserve the confocal filtering using one fibre. With these magnification and numerical aperture values, the axial resolution is of the order of 10 μm and the lateral resolution of the order of 1.4 μm . The numerical aperture is also chosen so as to optimize the number of 25 photons recovered, which must be as large as possible. The optical head can be made up of standard optics (doublet, triplet, aspheric) and/or index gradient (GRIN) lenses and/or diffractive lenses, presenting an optical quality and chromatics suited to confocality, i.e. minimizing the optical aberrations, which would otherwise conduct in particular to degradations over the

depth of field and consequently to the axial resolution of the apparatus. During operation, the optical head is intended to be placed in contact with the specimen 13. The latter is a biological tissue or a cell culture. The expression of the fluorescence is realized either by a fluorophore which is injected 5 (systemic fluorescence), or by a fluorophore produced by the cell itself by modification of a gene (transgenic fluorescence). In both cases, the fluorophore re-emits photons over a spectral band of greater or lesser width which can range from some ten nanometers to more than one hundred nanometers.

10 On the detection path, the fluorescence signal, exiting the rejection filter 8, is then focussed by means 9, made up for example of a detection lens, in a filtering hole of the spatial filtering means 10. The focal distance of the detection lens is calculated so that the fluorescence signal coming from a fibre is the same size as or slightly smaller than the filtering hole. The latter allows 15 the preservation of the fluorescence light coming only from the fibre illuminated by the incident beam. It allows rejection of the light which may have been coupled in the fibres adjacent to that which is illuminated. The size of the hole is calculated so that the image of a fibre fits into it perfectly. In this case, it is 20 μm . Moreover, still with a view to optimizing the quantity of 20 photons passing through the filtering hole, and thus the detected flux, the scanning means 4, the injection means 5, the focussing means 7 of the optical head, and the detection means 8, 9 and 10 are adapted to the detected fluorophore: these means are chosen to be sufficiently achromatic to collect photons over the broadest fluorophore emission band.

25 The detection means 11 have a maximum sensitivity at the fluorescence wavelengths studied. An avalanche photodiode (APD) or a photomultiplier can be used for example. Moreover, according to the invention, the pass-band is chosen in order to optimize the integration time of the fluorescence signal. It is 1.5 MHz, which

corresponds to the minimum sampling frequency of the image guide with an optimized integration time on each pixel.

The electronic and computer means 12 for control, analysis and digital processing of the detected signal and for display comprise the following cards:

5 - a synchronization card 20 which serves:

- to control in a synchronized way the scanning, i.e. the movement of the line M1 and frame M2 mirrors;

- to know at any moment the position of the laser spot thus scanned;
and

10 - to manage all the other cards via a microcontroller which can itself be piloted;

- a detector card 21 which comprises an analogue circuit which in particular carries out an impedance match, an analogue-to-digital converter then a programmable logic component (for example an FPGA circuit) which shapes
15 the signal;

- a digital acquisition card 22 adapted to the processing of a flux of digital data at variable frequency and its display on a screen 23;

- a graphics card 24.

By way of a variant, a single card can be used which combines the
20 functionalities of these different cards.

The image processing takes place in the following way.

When an image guide is placed in the apparatus, a first operation is carried out to recognize the design of the fibres in the image guide, and thus to learn the real location of each fibre intended to be used.

25 The following operations are also carried out before the apparatus is used:

- determination of the rate of injection particular to each fibre, using a homogeneous specimen, this rate of injection being able to vary from one fibre to another; and

30 - measurement of the background image, carried out without a specimen.

These two operations can be carried out regularly as a function of the frequency of use of the apparatus. The results obtained will be used for correction in operation of the digital signal on the exit of the detector card.

- 5 During operation, according to the invention 2 processing groups are carried out on the digital signal on the exit of the detector card:

The first group consists in a first step of correcting the digital signal in particular in order to take account of the real rate of injection particular to the fibre from which said signal has come and in order to subtract from it the part
10 of the flux corresponding to the background image. This allows processing of only a signal which really corresponds to the specimen observed. For this processing group a standard calculation algorithm is used which can be optimized in order to respect the constraint of real time.

- 15 The second group then comprises the reconstruction, from the corrected signal, of the digital image which will be displayed by the practician. The aim of the processing carried out is to provide for display a reconstituted digital image which is not simply a mosaic of image elements each corresponding to a corrected digital signal of a fibre placed side by side, but to provide a reconstituted digital image which no longer shows the fibres. An
20 algorithm is used for this which is intended to carry out a certain number of operations on each pixel, the algorithm being chosen so as to observe the constraint of real time, i.e. it must represent a good compromise between the complexity of the operations required, the quality of the result that can be obtained and the calculation time. By way of example, a Gaussian low-pass
25 filtering algorithm can be used.

- 30 The apparatus operates as follows. The source 1 produces a circular parallel excitation beam at a wavelength of 488 nm, which is then shaped in the afocal system 2 in order to give it the appropriate size for the best possible injection into the core of a fibre. This beam is then sent to the dichroic separation system 3 which reflects the excitation wavelength. The incident beam is then deflected angularly over time in two

directions in space by the optomechanical scanning system of mirrors 4, and injected using optical injection means 5 into one of the fibres of the image guide 6. The electronic means 12 serve to control the injection of one of the optical fibres of the image guide at a given moment angularly deflecting the
5 beam by means of the mirrors, point by point for a given line, and line after line, in order to constitute the image. On the exit of the guide, the light emerging from the injected fibre is focussed in the specimen by the optical head 7 at a point situated at a given depth situated between approximately some tens of μm and around one hundred μm . Thanks to the scanning, the
10 specimen is illuminated point by point. At each moment, the spot illuminating the tissue thus emits a fluorescence signal which has the particularity of being shifted towards greater wavelengths. This fluorescence signal is picked up by the optical head 7, then follows the reverse path of the excitation beam to the dichroic filter 3 which will transmit the fluorescence signal to the detection path. The interfering reflections which form at the excitation wavelength will
15 then be rejected by the rejection filter 8. Finally, the fluorescence signal is focussed in the filtering hole 10 in order to select only the light coming from the excited fibre and the photons are detected by the avalanche photodiode 11. The detected signal is then digitized and corrected. The detected signals,
20 one after the other, are processed in real time using the image processing described above in order to allow the reconstruction of an image in real time displayed on the screen.

According to the embodiment selected and represented in Figures 3 and 4, the apparatus comprises the same elements as the apparatus of
25 Figures 1 and 2 with the exception of the optical focussing head 7. The same reference numbers are used for these same elements in the figures and therefore will not be described again here in detail.

The image guide 6 used for this apparatus can be made up of approximately 5000 to 100 000 flexible optical fibres, depending on the
30 external diameter of the guide which is sought, which is itself a function of the intended application (endoscopy, desired field size, etc.). For example they may

be image guides marketed by FUJIKURA. The core diameter of the fibres is preferably between 1 and 4 μm . This leads to a beam divergence on the exit of the fibre presenting an angle of approximately 18° for a numerical aperture of 0.42 in the air. A core diameter of 1 μm can be obtained by a method of stretching the end of the entire image guide. The end of the guide is polished and does not contain optical means. The aim of the polishing is to give the surface of the guide, and thus the bare fibres in contact with the tissue, the same surface condition in order that on the one hand the obtained image background is as homogeneous as possible and, on the other hand, to suppress any problems of adherence of the fibres to the tissue, which would damage it. The polishing can be smooth or rounded in order to follow the shape of the tissue. The numerical aperture of each optical fibre is preferably chosen to be as large as possible in order to collect the maximum of fluorescence photons, i.e. for example 0.42. The intercore distance gives the lateral resolution of the image obtained (distance between two points of the image). The smaller this distance, the better the resolution will be, but this will be to the detriment of the size of the field to be imaged. A good compromise should thus be found between the number of fibres of the guide and the intercore distance between the fibres in order to obtain a good lateral resolution (between 2 and 8 μm) with a suitable field size for observing the tissue elements desired.

Two examples of guides which may be suitable according to the invention are given below:

		Example 1	Example 2
25	Number of fibres	6,000	30,000
	Diameter of the imaged field	300 μm	650 μm
30	External diameter of the guide	330 μm	750 μm
	Core diameter of a fibre	3 μm	1.9 μm

Intercore distance	4 µm	3.3 µm
Maximum integration depth of the signal from the surface (shown schematically by 5 the plane P in Figure 1) or axial resolution	15-20 µm	10-15 µm
Lateral resolution	4 µm	3.3 µm

During operation, the distal end of the image guide is brought into
10 contact with the specimen 13, the end surface of the fibres thus being directly
in contact with the surface of the tissue. The latter is a biological tissue or a
cell culture. The expression of the fluorescence is realized either by a
fluorophore which is injected (systemic fluorescence), or by a fluorophore
synthesized by the cell itself by modification of a gene (transgenic
15 fluorescence). In both of these cases, the fluorophore present in the
microvolume excited according to the invention re-emits photons over a
spectral band of greater or lesser width which can range from some ten
nanometers to more than one hundred nanometers.

The operation of the apparatus is the same as that described
20 previously with the following exception: on the exit of the guide, the divergent
light emerging from the injected fibre is diffused in the specimen and the
fluorescence signal is collected in a microvolume situated between the
surface and a maximum depth of 25 µm (depending on the core diameter of
the fibres and their NA). Thanks to the screening, the specimen is illuminated
25 microvolume by microvolume. At each moment, the microvolume excited in
the tissue thus emits a fluorescence signal which has the particularity of being
shifted towards greater wavelengths. This fluorescence signal is picked up by
the same optical fibre as the one which was used for the excitation, then
follows the reverse path of the excitation beam to the dichroic filter 3 which will
30 transmit the fluorescence signal to the detection path. The detected signals
are processed in real time, one after the other, using the same image
processing as that described above with reference to Figures 1 and 2
in order to allow the real-time reconstruction of an image displayed on the
screen.